NON-INVASIVE METHOD AND APPARATUS TO DETECT AND MONITOR EARLY MEDICAL SHOCK, AND RELATED CONDITIONS

FIELD OF THE INVENTION

This invention relates generally to the diagnosis of medical shock-related conditions and to instruments for this purpose. More particularly, the invention relates to methods and apparatus for the non-invasive detection of pre-shock, shock and shock-related conditions (other related causes of cardio-pulmonary distress), and for assisting in a patient's recovery from these conditions by monitoring changes in capillary flow in skin areas of peripheral body organs.

BACKGROUND OF THE INVENTION

The normal skin color at most sites on the human body is generally pink. Skin color depends on the amount of blood flowing in the capillaries through which blood flows from the arterioles to the venules. The present invention resides in non-invasive detection of hemodynamic changes in the skin arteriolar-capillary flow during states of pre-shock, shock and cardio-pulmonary distress. These changes are indicative of a reduction in blood delivery to an organ of the body.

Expressed in its simplest terms, shock is the consequence of an inadequate delivery of blood to a major organ of the human body. Unless shock is promptly treated, this deprivation of blood may give rise to a disturbance in the metabolism of the organ with a resultant damage thereto. Because of the serious consequences of shock, its detection and treatment is regarded medically as an emergency procedure in which time is of the essence.

Cellular damage to an organ may be reversed by prompt treatment of shock. But it is otherwise irreversible and may lead to the death of the patient. Recovery from shock therefore depends on the promptness of treatment. However, before a

patient can be treated for shock he must first be diagnosed to determine whether the patient is actually experiencing shock.

The treatment to be administered to a patient in shock depends on the nature of his condition. For example, for some shock conditions the appropriate treatment includes fluid resuscitation and the drug dopamine which acts to increase arterial perfusion pressure. Treatment for a shock condition must be administered with extreme care while the patient is being monitored.

A significant aspect of diagnostic instrumentation in accordance with the invention is that it is adapted to monitor as well as to detect shock-related conditions in a non-invasive manner. Using this instrumentation, one can make, even in a pre-hospital setting, an early diagnosis of shock as well as determine whether the drug being administered to a patient in shock is having the desired therapeutic effect.

Medical authorities classify shock syndrome in the following five categories:

(1) Hypovolemic shock

(2) Septic shock

(3) Cardiogenic shock

(4) Obstruction to cardiac filling shock

(5) Neurogenic shock

Hypovolemic shock, the most common type of shock, is caused by a massive loss of blood, plasma or fluid from the body of a patient, or the loss of fluid from an intravascular compartment. Such losses may be due to dehydration, vomiting, diarrhea, burns, or because of the abusive use of diuretics. A loss of blood and plasma is experienced in hemorrhagic shock such as in cases of blunt and penetrating trauma injuries, gastrointestinal bleeding, or Gynecologic/Obstetric bleeding. Many cases of bleeding are occult (e.g. slow internal bleeding), and therefore can not be diagnosed early.

Septic shock is caused by bacterial infection in which an endotoxin is released into the blood stream. The sequestration and pooling of blood in various vascular compartments reduces the availability of blood for the perfusion of other vital organs.

Cardiogenic shock is usually attributed to a massive myocardial infarction caused by extensive damage to the myocardium. This may be the result of arrhythmia in a patient suffering from heart disease. In this category of shock syndrome, the heart fails to pump properly, with a consequent reduction in arterial blood.

Obstruction to cardiac filling shock takes place when this filling activity is lessened or arrested by a massive pulmonary embolism, or by space-occupying lesions. Neurogenic shock results from a severe spinal cord injury, or from a massive intake of a depressant drug, causing a loss of vasometric tone.

The five categories of shock syndrome each represent other causes of cardio-pulmonary distress, or a shock-related condition. The term "shock-related condition", as used hereinafter, is intended to embrace all five categories.

The onset of a shock condition is characterized by the reduction in blood flow to skin tissue (decreased skin perfusion). This reduction in skin perfusion is the result of a profound vasoconstriction of the skin tissue arterioles, which leads to decreased capillary flow, and a resultant poor perfusion to the skin. In order to diagnose an early stage of shock, one must detect this early reduction in skin capillary flow. A useful clinical, bed-side test for poor skin perfusion is an estimation of Capillary Filling Time (CFT). When applying pressure onto a specific skin area, the capillaries below the depressed area collapse and blood is blanched therefrom, thereby causing the skin color in the depressed skin area to whiten. Upon abrupt release of this pressure, blood flows back into the capillaries and the original skin color is recovered.

CFT is defined as the time it takes for the original pink skin color to return after it had been blanched. In clinical practice, prolongation of the CFT for more than 2 second is considered a state of shock resulting from poor skin perfusion. This

well-known bed-side test, although subjective and inaccurate, is an important vital sign of a shock state. If an appropriate treatment has not been given early enough, the shock condition will continue to deteriorate, the arteriolar vasoconstriction will increase even further, as reflected by prolongation of the CFT, blood pressure will fall, and the patient may die. However, an appropriate prompt treatment at the early stage of shock will decrease vasoconstriction and shorten the CFT.

Known non-invasive methods to diagnose shock do not evaluate perfusion. These methods rely on the following cardiovascular parameters:

Blood pressure. An indirect parameter of shock. The measurement of blood pressure identifies shock only in its late stages when blood pressure drops (uncompensated shock).

Heart rate. An indirect parameter of shock. The specificity of this measurement is low because heart rate is also increased by other common physiological conditions, such as anxiety and pain.

The advantage gained by measuring the rate of blood perfusion by way of CFT instrumentation is that it enables early detection of a shock syndrome (compensated shock, prior to the reduction of blood pressure) and indicates its severity. This makes possible prompt treatment of patients who can then survive a shock-related condition which may be fatal if untreated or if treated too late.

Disclosed in US Patent 3,698,382 is an apparatus for measuring veno filling time which applies intermittent and uniform pressure to the skin of a patient. This instrument which measures capillary flow changes secondary to the compression of a vein comprises a light source for illuminating a skin area and photoelectric monitoring means sensitive to the coloration of the skin area. The instrument measures the rate at which color returns to the skin area after pressure thereon is released. However, there are major differences between the '382 apparatus and apparatus in accordance with the invention in that the former measures capillary flow changes resulting from mechanical pressure applied to a nearby vein and these changes in flow do not reflect a state of shock.

When measuring CFT it is essential that pressure be applied only to capillary vessels while maintaining normal blood flow. In a preferred embodiment of an apparatus in accordance with the invention, a programmable mechanical unit applies an accurate measurable amount of pressure to the skin.

In order to diagnose the condition of shock, one must detect capillary flow changes resulting from the physiologic stress of shock. These changes in capillary flow are due to vasoconstriction and are not related to mechanical pressure applied to a near by vein. When measuring CFT, it is vital that pressure be applied only to the capillary vessels while maintaining normal venous flow. In contradistinction to the apparatus in the '382 patent, an apparatus in accordance with the invention uses a programmable mechanical unit that applies accurate measurable pressure to the skin, which increases gradually, until a point of maximal skin whitening has been detected. This technique makes it possible to find the MINIMAL blanching pressure which results in maximal whitening. At minimal blanching pressure, blood is moved away from the capillaries while maintaining normal flow in the veins. This technique is the hallmark of measuring true systemic changes in capillary flow.

The '382 patent apparatus is subject to interference from external light sources and therefore requires an opaque housing for the monitoring apparatus. The apparatus does not measure skin temperature which has an independent effect on capillary flow. In addition, the mechanical arrangement required for maintaining uniform pressure in order to attain more accurate readings is cumbersome and costly.

They are also relatively complex and expensive and difficult to interpret clinically (laser Doppler devices for example). Time is of the essence in the diagnosis and treatment of shock, yet known types of skin capillary flow instrumentation are incapable of facilitating rapid diagnosis and treatment of shock. It is vital that skin capillary flow instruments have a high order of accuracy so that their readings indicate the severity of the shock or shock-related condition.

Studies published in the medical literature over the last two years demonstrate that skin temperature independently influences the skin capillary flow. One major limitation of prior skin capillary flow measurement devices is that they do not take into account skin temperature, and therefore do not correlate the measurement to skin temperature. This correlation enables real-time analysis of the state of shock. In contradistinction, a device in accordance with the invention measures skin temperature prior to each CFT measurement, so that every CFT measurement is correlated to the change in skin temperature.

Of general background interest is US Patent 4,494,550 which discloses apparatus for the non-invasive detection of venous and arterial blood flow drainage disorders which is designed for the detection of flow abnormalities in the large vessels of a limb. Also of background interest is US patent 5,050,613 (1991) which discloses a vascular testing apparatus. This includes capillary blood flow sensors to measure the blood flow of a patient. This diagnostic tool acts to determine the overall vascular integrity of a patient, but is unable and does not diagnose shock or shock-related conditions.

SUMMARY OF THE INVENTION

In view of the foregoing, the main object of this invention is to provide a diagnostic method and an instrument for carrying out the method to determine accurately whether a patient is suffering from a state of shock and shock-related conditions, as well as to measure and monitor the severity of this physiologic condition.

In particular, an object of this invention is to provide a non-invasive method and apparatus adapted to detect pre-shock, shock and shock-related conditions by ongoing measurements of the patient's capillary filling time (CFT).

A significant advantage of an apparatus in accordance with the invention is that it can expedite recovery by monitoring changes in capillary flow in skin areas of peripheral body organs. The CFT measuring instrument provides a rapid yet

accurate reading of the patient's condition, making it possible to treat the patient without delay to avoid damaging consequences.

It is also an object of this invention to provide a CFT diagnostic instrument which is of relatively simple design and easy to operate.

Briefly stated, these objects are attained in a diagnostic medical instrument adapted to determine whether a patient is suffering from a pre-shock, shock, or shock-related condition. Some shock-related conditions are related to inadequate flow in a specific organ. These medical conditions are common in patients after orthopedic surgery, flap reconstruction surgery, or patients who suffer from a severe peripheral vascular disease. By being highly sensitive to changes in capillary flow, an apparatus in accordance with the invention is applicable to these medical shock-related conditions.

The instrument is used in a capillary filling time test procedure in which a skin area in the patient overlying blood-filled capillaries which normally display a pink color, is depressed to expel blood from the capillaries and to blanch the skin and impart a white color thereto. When a point of blanching has been attained at a minimal pressure point, the pressure is then released to permit blood to flow back into the capillaries and cause the skin to regain its natural pink color. Using this minimal blanching pressure technique, blood is withdrawn from the capillaries whereas venous blood flow remains almost intact.

The instrument includes a color sensor trained on the skin area and responsive to light reflected therefrom to produce a first signal at the point in time the depressed skin color is blanched from pink to white and pressure is released when blanching at minimal pressure is attained, to later produce a second signal at the point in time at which the skin color regains its natural pink color. Herein, "color sensor" refers to any light sensor capable of sensing intensities of light within any desired range of wavelengths, for example the full range of visible light, or any other range of wavelengths, either within the visible range, beyond the same or overlapping both, among others. When the post-blanching skin color corresponds to a pre-test natural color, the CFT can be detected by recording the

time which has elapsed from the maximal blanching point to this final point. In other words, the time elapsing between the first signal (starting point of minimal blanching pressure release) and the second signal (final point where post-blanching color equals pre-test color) is measured to provide a CFT index indicative of the patient's condition at the time the test was conducted.

For each pre-determined time interval, this measurement is repeated and a new CFT is recorded.

The device will continue measuring CFT at any desired interval, for example every 30 seconds to 1-5 minutes (this depends on clinical demands), and a change of CFT over time will be recorded and monitored. This change in CFT, or d[CFT]/d[t], reflects skin perfusion changes over time and measures deterioration or improvement of shock state.

In one preferred embodiment of the invention, the color sensor includes a video camera trained on the skin area of the patient and responsive to light reflected from this area to yield an image signal whose character depends on the existing color of the skin.

In another embodiment, the skin area is illuminated by a beam of light modulated at a predetermined frequency, the pulsed light reflected from this area being intercepted by a photosensor whose output signal is indicative of the skin color. In yet another embodiment, the skin area is illuminated by non-modulated light.

Optionally, the CFT may be corrected for distance effects introduced by the displacement of the skin during spring-back from the depressed position during CFT testing. Alternatively, the apparatus may be configured to minimize such distance effects.

Optionally, the CFT may be adjusted to take account of the temperature of the patient. The relationship between CFT and temperature may be determined empirically, using statistical tools.

Further, heating effects due to the apparatus itself may also be compensated for.

Optionally, potentially false color readings originating from capillary damage due to repeated testing of a skin area may be avoided by sensing the color changes in an area close to but not including the area of skin that is being directly pressured by the apparatus of the invention.

BRIEF DESCRIPTION OF THE DRAWINGS

In order to understand the invention and to see how it may be carried out in practice, a preferred embodiment will now be described, by way of non-limiting example only, with reference to the accompanying drawings, in which:

Fig. 1A illustrates the structure of a skin color sensing apparatus for the diagnosis of a shock-related condition in a patient by measuring the capillary filling time and rate in accordance with a first embodiment of the invention;

Fig. 1B schematically illustrates the color sensor included in the apparatus shown in Fig. 1;

Fig. 2 is a block diagram of the apparatus shown in Fig. 1 for the diagnosis of a shock-related condition in a patient by measuring capillary filling time and rate;

Fig. 3A is a graphical representation of the measurement CFT results;

Fig. 3B is a graphical representation of CFT, as a function of the level of shock, for obtaining inferences related to the trend of the patient's physiological condition in reaction to medical treatment;

Fig. 4 schematically illustrates how the apparatus is used, as shown in Fig. 2, for the diagnosis of pre-shock state in a patient;

Fig. 5 illustrates the color sensor included in a second embodiment of a CFT diagnostic instrument;

Fig. 6 is a block diagram of the apparatus included in the second embodiment;

Fig. 7 is a graph showing the relationship between CFT readings and conditions of shock;

Fig. 8 is a graph showing the relationship of skin temperature on CFT readings;

Fig. 9 is a block diagram of the apparatus according to a third embodiment.

Fig. 10 illustrates an embodiment of the invention according to a second aspect thereof.

DETAILED DESCRIPTION

First Embodiment: Schematically illustrated in Fig. 1A is a CFT instrument 450 adapted to diagnose a shock-related condition in a patient by measuring capillary filling time and rate.

Instrument 450 includes a camera 412, such as a color video camera, fixed in place by a holder 414 above a rigid table surface 411 on which an appendage 410 of a patient rests. This appendage may for example be the patient's finger. The position of camera 412 is adjusted so that the skin area 413 viewed by the camera for purposes of CFT measurement, is in or is close to the focal plane of the lens. Pressure may be applied to and released from skin area 413 manually or by using mechanical apparatus which may be automatically controlled.

Skin area 413 is illuminated by background light, and light reflected from the surface of this area is received in the lens of camera 412. A minimal illumination level of 0.2 lux is sufficient for most currently-available modern cameras for color discrimination. Camera 412 generates an electrical signal having a magnitude corresponding to the particular color of the image received by the camera, this signal being fed by a line to a processing and display unit 400. In the event the illumination level of the background light is insufficient, skin area 413 may be illuminated with a light source, such as a conventional lamp or a Light Emitting Diode (LED).

A sensing device 100 as shown in Fig. 1B, is connected to the processing and display unit 400 by an electrical cord through which CFT data is fed for processing and display. The processing and display unit 400 may be a personal

computer that uses control and processing software to process the data received by the lens of camera 412, and calculate the CFT total time and rate. Pressure is applied and released manually by the user in accordance with instructions provided by processing and display unit 400. The processing and display unit 400 may further include circuitry for controlling automated application of pressure.

The control circuitry may also be used to select a specific area for processing taken from the imaged skin area. Such selection may be carried out, for example, by software which controls the processing. The sensing device may also be attached to other locations in the patient's body that are rich in subcutaneous blood vessels, such as to the lip or to the ear lobe, for measuring the CFT.

Fig. 1B schematically illustrates the structure of a skin color sensing device 100 for the diagnosis of a shock-related state in a patient by measuring the capillary filling time and rate. Device 100 comprises a camera 412, such as a color video camera, contained in a transparent external housing 102, whereby most of the background light enters through this external housing and illuminates the skin surface 106.

Device 100 may further include an optional light source 101, such as an LED, operated by a power supply during measurement when background illumination is not at a level sufficient to enable the camera 412 to discriminate between colors. External housing 102 may be light reflecting with an opening in its bottom side, so that most of the light radiation emitted from light source 101 is directed toward the bottom side in one direction "A".

External housing 102 may also include an opaque internal housing 104, having an opening in its bottom side, so as to enable light radiation to enter into the opaque internal housing space only from its bottom side. Using this structure, camera 412 in internal housing 104 receives most of the light reflected from the skin. The bottom sides of external housing 102 and internal housing 104 are aligned with each other and covered by a transparent rigid layer 105. Layer 105 acts to apply pressure on the skin while enabling light to pass through in both directions.

Transparent rigid layer 105 is brought into contact with an exterior layer 106 of the skin of the patient being diagnosed. Pressure is applied manually or automatically on the external housing 102 toward the skin surface in a perpendicular direction A. External housing 102 delivers the pressure to the transparent rigid layer 105, which transfers it through exterior layer 106 to the interior layer 107 of the skin containing most of the subcutaneous blood vessels (capillaries). When the magnitude of applied pressure is adequate for expelling blood from the capillaries and maintained for a sufficient period of time, blood is forced out of the capillaries and the color of the interior layer 107 of the skin becomes much brighter (i.e. close to white).

The background light as well as light radiation emitted from light source 101 penetrates the skin and is partially reflected back in direction "B" into internal housing 104. The degree of reflection from interior layer 107 is inversely related to the blood flow in the capillaries under pressure inasmuch as blood absorbs light, the more blood the less the amount of reflected light. The reflected light enters the lens of camera 412, which produces an electric signal whose magnitude depends on the instantaneous color of the skin. The position of camera 412 within the device 100 is arranged so that the exterior surface of the transparent rigid layer 105 is essentially in the focal plane of the camera 412. This positioning results in a clear and focused image that is received by the camera lens. A focused image sharpens the distinction between colors and therefore enhances the resolution and accuracy of the measurement.

Under zero pressure (i.e., full blood flow), a patient's skin color is normally pink, and less light radiation is reflected back from the capillaries. When the skin is subjected to a pressure to arrest blood flow, the skin color then becomes white and more light radiation is reflected back from the capillaries. Therefore, changes in magnitude of the electric signal yielded by camera 412 afford an accurate index to capillary filling time and rate which commences upon releasing the pressure from the skin. Device 100 is connected to a power supply for operating the optional light source 101 and for operating data collection, processing and display circuitry for

processing the signals provided by the camera 412 and displaying the measurement results.

Fig. 2 is a block diagram of an apparatus 200 for the diagnosis of a shock-related state in a patient by measuring capillary filling time and rate in accordance with the invention. Apparatus 200 includes camera 412, whose output is supplied to a frame grabber 206 for capturing the image received by the camera. Light reflected from the skin surface is converted by camera 412 to a corresponding video signal, such as a Composite Video or a Red-Green-Blue (RGB) Video signal, depending on the type of camera used, that represents the received image.

The video signal is fed into an electronic circuit (e.g., a Frame Grabber or a Video Capture circuit) which decodes the video signal and converts it into a corresponding array of digital values, which array is stored in a memory. Each cell of the memory stores a digital value that represents the light intensity and the color of a portion of the received image. Camera 412 updates the image at a rate of 50 times per second, and therefore, the image information, generated by frame grabber 206 and stored in the memory array is also updated at the same rate. A rate of 50 times per second usually corresponds to video cameras compatible with Pulse Alteration by Line (PAL) video encoding standards. A rate of 60 times per second usually corresponds to video cameras compatible with National Television System Committee (NTSC) video standards. Faster video cameras to update the image at higher rates are also useable.

The digital data is fed into a digital processor 207 which analyzes the data and display the results on display 208. Processor 207 samples a desired area of the image which contains most of the tested skin area. At the next step, processor 207 calculates the intensity of the essentially pink/red light, reflected by the tested skin area. The intensity of the reflected light is processed and normalized to a baseline, which may be the normal color of the patient's skin when no pressure is applied. The image information is updated in a rate determined by the type of camera included in the system. Processor 207 therefore continuously calculates the normalized intensity.

Display 208 presents a display of the calculated results of the normalized intensity (i.e., the CFT) as well as a graphical representation of the measurement process as a function of time. The graphical representation indicates whether or not the measurement results are reasonable, and if desired, the measurement can be repeated. Other data processed results, such as statistical data, can be also displayed to provide indications regarding the reaction of the patient to medical treatment.

Fig. 3A is a graphical representation of CFT measurement results. At the first stage, no pressure is applied on the skin, and therefore the apparatus 200 can carry out calibration of the initial skin color of the patient. The value of the calibration is stored for use at the end of the measurement. The calibration process is essential in that the normal color of the skin depends on the individual and differs from patient to patient.

At the second stage of operation, pressure is applied to the skin at a magnitude and for a duration sufficient to obtain maximum whitening of the skin color in the depressed area. The processor can be programmed to provide a warning signal (such as a beep) to the user when the pressure is insufficient or shorter in duration than required. Obtaining maximum whitening of all the depressed area is indicative of sufficient whitening pressure.

Stronger pressures of longer duration do not affect the skin color beyond maximum whitening. After obtaining maximum whitening, a signal indicative thereof is provided to the user to quickly release the pressure. Measurement of the CFT is started at that instant (to) at which the skin coloring proceeds to change from its maximum whitening color to regain its original pinkish color. Normally, the rate of filling is higher at the beginning of the filling process and lower as time lapses.

The apparatus uses the stored calibration value to determine the moment tf at which the normal pink skin color is regained, at which point the measurement ceases. The recovery time can be determined by the desired degree of measurement accuracy. For example, point tf can be defined as the instant at which the value of the digital word that corresponds to the current skin color reaches a value that is

90% of the value of the digital word that corresponds to the original skin color of the patient being diagnosed. In the graph of Fig. 3A, the CFT reading is given by tf – to.

The accuracy of the measurement can also be determined by the rate of change in the skin coloring in the time interval that is close to the conclusion of the measurement. The last segment of the graph lies between the points of time t1 and tf. The rate of change in this time interval is nearly constant and is nearly insensitive to the magnitude and duration of the applied pressure. Hence, the CFT can be extrapolated with relatively high accuracy from the time interval tf – t1. Under normal conditions CFT should be below one second. A CFT value above two seconds can be regarded as representing a pre-shock state. Longer CFT values can be considered to be indicative of more severe shock states.

Fig. 3B is a graphical representation of the CFT as a function of shock-state for obtaining inferences related to the trend of the patient's physiological condition in response to medical treatment. In the initial time interval between time-points t2 and t3, the CFT value is then below 2 seconds, hence the patient is in a normal, shock-free condition. An early and mild shock condition starts at time-point t3 where the CFT value exceeds 2 seconds. As time lapses with no proper treatment of the shock condition, the shock becomes more severe until time-point t4 is reached. This point indicates the entry of the patient into a moderate shock condition (CFT value higher than 3 seconds). The next stage is indicated by the time-point t5. This indicates the entry of the patient into a late (severe) shock condition (CFT value higher than 4 seconds). From point t5 and beyond, the CFT rises rapidly.

Analysis of skin temperature is crucial for the clinician to make an appropriate diagnosis and monitoring of shock. For example, very cold skin temperature will independently prolong CFT (an acceptable false positive of CFT measurement). For each time interval, the device will measure and monitor both CFT and skin temperature. (See "Modified Second Embodiment" in connection with Fig. 6).

When a medical treatment is administered to the patient, the CFT is measured thereafter on a periodic basis. This measurement is used to determine whether the pre-shock or the actual shock condition is reversible. If the patient's reaction to the given treatment is positive, then in time the CFT will be reduced, indicating a significant improvement in the physiological condition of the patient until the CFT value goes below the safe 2 Sec level.

Fig. 4 schematically illustrates the use of an apparatus 200 for the diagnosis of pre-shock state in a patient. Apparatus 200 includes a skin color sensing device 100 attached by straps or by adhesive tape to a skin area rich in subcutaneous blood vessels, such as hand fingers, and a processing and display unit 400 coupled to sensing device 100. Device 100 is connected to the processing and display unit 400 by an electrical cord through which the CFT data is fed for processing and display. Pressure is applied and released manually by the user in accordance with instructions provided by processing and display unit 400. The sensing device for measuring CFT may also be coupled to other sites in the patient's body that are rich in subcutaneous blood vessels, such as to the lip or to the ear lobe.

In practice, an automatic measurement can be carried out by integrating a mechanical control apparatus into sensing device 100 adapted to control the applied pressure and release thereof by an external controller. Such mechanical apparatus may comprise a miniature linear motor that produces linear movement rather than rotational movement. Alternatively, linear movement pressure can be applied by an electromagnet or by a rotational motor with an eccentric movement mechanism. The linear movement can be controlled to depress a movable member, such as a movable transparent rigid layer, against the skin and to release the pressure by a corresponding control command.

Sensing device 100 and the processing and display unit 400 may further include receiving and transmitting circuits to enable wireless exchange of data and control commands required for CFT measurements. Wireless connection makes feasible a single processing and display unit 400 to control and monitor several sensing devices 100, each attached to a different patient. Each sensing devices 100

is identified by a unique code assigned to it, to eliminate false associations between processed data and a patient..

The invention can be carried out in a great variety of other ways, employing techniques which differ from those described herein, such as by using pneumatic apparatus for applying pressure to the patient's skin, or by using an Infra-Red camera rather than a video camera. Also one can store the history of CFT measurements of a patient and display the variation of the CFT curve with time.

Second Embodiment: This embodiment of a CFT diagnostic instrument differs from the instrument shown in Fig. 1 mainly in the nature of its skin color sensor. However, in all other respects it operates in the same manner as does the first embodiment.

Fig. 5. Schematically illustrates the structure of a skin color sensing device 500 for the diagnosis of a shock-related state in a patient by measure the capillary filling time and rate according to the second embodiment of the invention. Device 500 includes a continuous (non-modulated) or a pulsating (modulated) light source 501, such as a Light Emitting Diode (LED) driven by a rectangular voltage pulse generator at a predetermined frequency fo. Light source 501 is enclosed in a light-reflecting external housing 502 having an opening in its bottom side so that most of the light radiation emitted from light source 501 is directed toward the bottom side in one direction "A". External housing 502 has within it an opaque internal housing 504 containing a light sensor 503, such as a photodiode, a phototransistor, a photo-resistor or a photoelectric cell. Internal housing 504 has an opening in its bottom side which permits light rays to enter therein only through its bottom side. The bottom sides of external housing 502 and internal housing 504 are aligned with each other and are covered by a transparent rigid layer 505. This layer serves to apply pressure on the skin while enabling light to pass therethrough in both directions.

Transparent rigid layer 505 of device 500 is pressed into contact with the exterior layer 506 of the skin. Pressure is applied manually or automatically on the external housing 502 toward the skin surface in a perpendicular direction. The

external housing delivers the pressure to the transparent rigid layer 505 which transfers it through exterior layer 506 to the interior layer 507 of the skin containing most of the subcutaneous blood vessels (capillaries).

As a result, when the magnitude of the applied pressure is adequate and is maintained for sufficient period of time, blood is then forced out of the pressurized capillaries and the color of the interior layer 507 of skin becomes much brighter (i.e. substantially white). Light rays emitted from light source 501 penetrate into the skin and are partially reflected back in direction "B", into the internal housing 504. The degree of reflection from interior layer 507 is inversely related to blood flow in the capillaries under pressure inasmuch as blood absorbs light, the more blood in the capillaries the lesser is the reflected light.

The reflected light is aggregated by light sensor 503 which yields an electric signal whose magnitude depends on the instantaneous color of the skin. Under zero pressure (i.e., full blood flow), the skin color is normally pink and therefore less light is reflected back from the capillaries. When the skin is subjected to pressure and blood is expelled from the capillaries, the skin color is then white. Hence when the skin is pink, the intensity of reflected light is relatively low and when the skin is white the intensity of reflected light is significantly higher. Consequently, changes in magnitude of the electric signal produced by light sensor 503 affords an accurate measure of the capillary filling time and rate. The Device 500 is connected to a pulsed power supply for energizing light source 501 and for operating data collection, processing and display circuitry to process the signals yielded by light sensor 503 and for displaying the measurement results.

Fig. 6 is a block diagram of an apparatus 600 in the second embodiment for diagnosing a shock-related state in a patient by measuring capillary filling time and rate. Apparatus 600 comprises a rectangular pulse oscillator 601 operated at a suitable frequency, for example fo = 18 KHz. The output of oscillator 601 is fed into a driver 602 which provides rectangular output pulses having sufficient energy to power light source 601 to emit light pulses at the same frequency fo. Light reflected from the skin is converted by light sensor 603 to a corresponding

pulsatory electrical signal. This signal is fed into an amplifier 604 operating within a frequency band that includes frequency fo to increase the amplitude of the electrical signal.

Light sensor 603 is may be sensitive to the full color spectrum, or alternatively most sensitive to light radiation to a particular range of wavelengths, for example between red and infra-red in the color spectrum to a particular range of wavelengths, for example also to background light sources, such as external light radiation which adds an unwanted 50/60 Hz signal, or to sunlight which adds an unwanted DC level. Therefore the electrical output signal includes interfering components as well as the desired component at frequency fo. The interfering components are reduced in magnitude by the amplifier 604 which is tuned to amplify the desired component at frequency fo to a greater degree than the unwanted components.

The amplified electrical signal from amplifier 604 is further filtered by a Band-Pass-Filter (BPF) 605. This filter is tuned to pass only the desired component at frequency fo and to reject all other unwanted components. BPF 605 is implemented as an active filter using Integrated Circuit (IC) technology. The resultant filtered signal at the output of BPF 605 is a rectified sine wave which is fed into an integrator circuit 606. Integrator circuit 606 outputs a Direct Current (DC) level proportional to the magnitude of the rectified sine wave and hence the magnitude of light reflected from the skin. It is therefore highly sensitive to changes in skin color.

The DC signal is fed into an Analog to Digital Converter (ADC) 607, which converts the DC level into a corresponding digital word. The digital data is fed into a digital processor 608 which analyzes the data and display the results on a suitable display 609. Display 608 exhibits a digital value representing the measurement results (i.e., the CFT), and a graphical representation of the measurement process as a function of time. The graphical representation provides an indication of whether or not the measurement results are reasonable, and if desired, the measurement can be repeated. Other data processed results, such as statistical data, can be also

displayed to provide indications related to the reaction of the patient to medical treatment.

Third Embodiment: The third embodiment is substantially similar to the second embodiment, as described herein, with the following differences, mutatis mutandis. Fig. 9 is a block diagram of an apparatus 700 in the third embodiment for diagnosing a shock-related state in a patient by measuring capillary filling time and rate. Apparatus 700 comprises a constant source 712 operated at a DC voltage. The output of source 712 is fed into a driver 702 which provides energy to power light source 701 to emit non-modulated, continuous light. Light reflected from the skin is converted by light sensor 703 to a corresponding electrical signal. This signal is fed into an amplifier 704 operating at near- DC frequency band to increase the amplitude of the electrical signal.

Light sensor 703 is may be sensitive to the full color spectrum, or alternatively most sensitive to light radiation to a particular range of wavelengths, for example between red and infra-red in the color spectrum The sensor 703 may also to background light sources, such as external light radiation which adds an unwanted 50/60 Hz signal, or to sunlight which adds an unwanted DC level. Therefore the electrical output signal may include interfering components as well as the desired DC level. The interfering components are reduced in magnitude by the amplifier 704 which is tuned to amplify the desired DC signal to a greater degree than the unwanted components.

The amplified electrical signal from amplifier 704 is further filtered by a Low-Pass-Filter (LPF) 713. This filter is tuned to pass only the desired component of low signal frequencies and to reject all other unwanted components. LPF 713 is implemented as an active filter using Integrated Circuit (IC) technology. The resultant filtered signal at the output of LPF 713 is a direct current (DC) level proportional to the magnitude of the light reflected from the skin. It is therefore highly sensitive to changes in skin color.

The DC signal is fed into an Analog to Digital Converter (ADC) 707, which converts the DC level into a corresponding digital word. The digital data is fed into

a digital processor 708 which analyzes the data and display the results on a suitable display 709. Display 709 exhibits a digital value representing the measurement results (i.e., the CFT), and a graphical representation of the measurement process as a function of time. The graphical representation provides an indication of whether or not the measurement results are reasonable, and if desired, the measurement can be repeated. Other data processed results, such as statistical data, can be also displayed to provide indications related to the reaction of the patient to medical treatment.

Fig. 3A which is a graphical representation of the measurement results of the CFT obtained with the first embodiment of the invention is also representative of the results obtained with the second embodiment. At the first stage, no pressure is applied on the skin, and therefore the diagnostic apparatus can carry out calibration of the initial skin color of the patient which is a shade of pink.

The calibration value is stored for use at the conclusion of the measurement. The calibration process is essential, since the normal color of the skin depends on the individual being tested and differs somewhat from patient to patient. At the second stage, pressure is applied with a magnitude and duration sufficient to obtain maximum whitening of the skin color in the depressed area. The processor can be programmed to provide a warning signal (such as a beep) to the user, that the pressure is not sufficient or is shorter in duration than required. Obtaining maximum whitening of the entire depressed area is indicative of sufficient pressure.

After obtaining maximum whitening, a corresponding signal is provided instructing the user to quickly release the pressure. Measurement of the CFT is initialed at that moment, "to". The skin coloring proceeds to change from maximum whitening to essentially the original pinkish color. Normally, the rate of filling is higher at the beginning of the filling process and lower as time lapses. The apparatus uses the stored calibration value to determine the moment tf, at which the original skin color is recovered and the measurement terminated. Recovery time can be determined in accordance with the desired measurement accuracy. For

example, tf can be defined as the instant at which the value of the digital word that corresponds to the current skin color reaches a value which is 90% of the value of the digital word that corresponds to the original skin color of the patient being tested. In the graph of Fig. 3A, the CFT is given by tf – to.

The signal representative of changes in skin coloring can also be affected by optical amplitude variations, which may be caused at times by the movement of skin back to its original position after the pressure is released by the sensor, for example. In order to correct for this effect, the processing procedure for the signals may be modified to include a compensating algorithm that may be applied before the computation of CFT time.

The accuracy of the measurement can also be determined by the rate of change in the skin coloring, in the time interval that is close to the completion of the measurement. The last segment of the graph appears between the time points t1 and tf. The rate of change in this time interval is nearly constant, and is almost insensitive to the magnitude and duration of the applied pressure. Hence the CFT can be extrapolated with relative accuracy from the time interval tf – t1.

The CFT under normal shock-free conditions should be below 1 Sec. When a CFT value rising above 2 Sec is diagnosed. This is indicative of a pre-shock state. Longer CFT values indicate a more severe shock condition.

Fig. 3B which is a graphical representation of the CFT in the first embodiment for obtaining inferences related to the trend of the patient's physiological condition in reaction to medical treatment, is also applicable to the second embodiment.

Modified Second and Third Embodiments: The color sensor included in the second embodiment of CFT diagnostic apparatus does not take into account the temperature of the patient's skin at the time of the diagnosis and its effect on the CFT reading. For accurate readings it is necessary to measure the skin surface temperature and record it prior to each CFT measurement.

In order to factor into the processing of the reflected light intensity the influence thereon of skin temperature, included in the color sensor shown in Fig. 6

for the second embodiment or in Fig. 9 for the third embodiment is a heat sensor 610, such as an infrared detector or a thermistor, whose output signal varies in magnitude as a function of the intensity of infrared rays emanating from the skin surface in the course of CFT diagnosis. Infrared detector 610 is responsive only to the heat of the skin, not to light reflected from the skin surface.

The electrical signal yielded by heat sensor 610 is not pulsed and has a magnitude which is a function of skin temperature. This signal is digitized in an A/D converter 611 whose digital output is entered into computer microprocessor 608. Microprocessor 608 is programmed by software to factor into the CFT reading the effect thereon of skin temperature. This corrected reading is of value in real time diagnosis of a patient's shock-related state, for it takes into account the skin temperature of the patient when in shock. It is of somewhat lesser value when monitoring the condition of a patient being treated for shock.

A preferred form of skin temperature sensor is a thermometer which can be placed directly on the skin surface of a patient being diagnosed for shock, to provide an electrical signal whose magnitude depends on the existing skin temperature. The thermometer signal is entered into microprocessor 608 of a computer into which is also entered the CFT signal indicative in terms of seconds, the shock state of the patient.

Fig. 8 illustrates the effect of skin temperature on CFT readings for patients 1 and 2 having different skin temperatures T1 and T2, where T2 is greater than T1. It will be seen that in a normal no-shock state, the CFT readings which indicate this state in terms of seconds are different, thereby reflecting the effect on the CFT readings of the degree of difference between temperatures T1 and T2. Similar differences appear for the pre-shock and shock states.

Especially for the second and third embodiments, and for all embodiments where the distance between the color or light sensor and skin is small, the depression of the skin under the action of the mechanical pressure inducer (eg a plunger) may have an influence on the intensity of light finally reaching the

sensor. This is so when the amplitude of the skin depression is not insignificant with respect to the color or light sensor-to-skin distance. When the mechanical pressure inducer is at maximum depth with respect to the skin or tissue, the distance to the sensor is greater, and thus intensity of the light received by the sensor is lower, in line with the inverse square law. When the skin springs back, after the mechanical pressure is released, i.e., at the beginning of the measurements for CFT, the distance progressively reduces, and the intensity progressively increases. Thus a positive intensity effect occurs during the monitoring of the skin color or light intensity after blanching due to the skin returning to its original position. At the same time, there also occurs a negative intensity effect, i.e. a falling in the intensity measured by the color sensor, due to the color of the skin changing from white to pink. While the sensor senses the combined effect of positive and negative effect, it is only the negative effect due to CFT that is of interest. According to another aspect of the present invention, the intensity effects due to distance may be corrected or eliminated at source to obtain the true changes in intensity due to changes in color.

In one embodiment of the invention, the intensity effects due to changes in distance are compensated by first determining the spring-back properties of the skin when the mechanical pressure is released. Knowledge of these properties enables the changes in distance with respect to time for the skin to be calculated during the restoration period, as the skin returns to the original position. The variation of distance with time can in turn be converted into relative changes in intensity, since the intensity obeys an inverse square law with respect to distance. The relative changes in intensity can then be related to a baseline intensity value, such as the original intensity that is recorded just after the mechanical pressure is released, for example. Alternatively, the baseline intensity may be the original intensity of the illuminating radiation, i.e., the intensity at the source, in which case the intensity is inversely proportional to a 4^{th} power of the distance. These spring-back properties of the skin may change from patient to patient, and from

device to device, and may also vary even with the same patient, form example depending on the degree of dehydration of the patient.

Considering the skin (or other tissue) to behave as a spring model, the resistance of the skin to deformation by the mechanical pressure inducer may be assumed to be in some way proportional to the depth of the pressure inducer with respect to the skin. Suitable stress or strain measurement means may be provided, together with displacement measurement means, and thus the spring constant (which may actually vary with depth) of the skin under the particular conditions of the current CFT test may be obtained. Once the inducer is released from the skin, a suitable algorithm can estimate the trajectory of the skin back to the original position using the established spring constant, and thus the changes in distance with time for the skin can be converted to an intensity effect. This intensity effect may then be subtracted from the actual intensity recorded via the color or light sensor to provide a corrected intensity value for the light received from the skin or tissue being tested which is indicative of CFT effects.

In another embodiment of the invention, the distance between the skin or tissue being tested and the color or light sensor is kept constant during capillary filing, such that no substantial spring-back occurs. Referring to Fig. 10, for example, the device 800 may comprise a guard 810 in the form of a ring 815 that is spaced from the body 850 of the device via struts 820. A mechanical plunger 830 moves fro a retracted position, displaced from the ring 815, to a deployed position just below the level of the ring such as to provide pressure to the skin. As the plunger is retracted, the pressure is released from the skin but this is prevented from springing back due to the ring. The body contains the color or light sensor (not shown), as well as other components such as illumination means, for example.

As has been described above, a temperature sensor may be used to determine skin temperature, which can then be used to correct the CFT for temperature effects.

It is important to realize that it is desirable to determine the CFT of a patient at the actual skin temperature of the patient that is not influenced by the device of the invention itself. Typically, skin temperature should be a function of the internal perfusion effects in the skin. However, the closeness of the device, to the skin, particularly when taped thereto generates some local warmth, as the part of the skin covered by the device is now at least partially insulated from the outside environment. In addition, the illumination source itself can also generate some additional warmth to the skin, the temperature of which naturally increases. Preferably, and as illustrated in Figs 6 and 9, a heat sensor 610 may be provided outside the main body of the device, and substantially beyond the influence of the illumination source or the main contact point between the device and the skin. This heat sensor thus provides a skin temperature Ta, and at the beginning of testing, the part of the skin being tested is at this temperature. As testing continues, this part of the skin gets progressively warmer, until steady state conditions are reached, wherein the temperature of this part of the skin reaches Tb, higher than Ta. At such conditions, the CFT determined with respect to the skin portion is thus associated with Tb rather than Ta, and needs to be corrected to Ta, which is more representative of the skin temperature minus the device temperature effects. According to this aspect of the invention, a second temperature sensor is provided for measuring the temperature of the skin, substantially similar to sensor 610 as described herein, mutatis mutandis, but such that it is influenced by the heating effects of the illumination means and the main contact points between the device and the skin. Thus, referring to Fig. 5, the second temperature sensor (not shown) may be located next to the light sensor 503 within internal housing 504, while the first sensor (not shown) may be provided outside of the external housing 502. According to this aspect of the invention, the temperatures Ta and Tb are measured via the first and second heat sensors, respectively, and suitable processing means monitors the changes in temperature as a function of time. At the beginning of testing, when Tb is increasing with respect to Ta, the CFT measurement may be adjusted according

to temperature Ta. As the skin portion being monitored warms up due to the closeness of the probe, and due to heating from the light source, the CFT eventually corresponds to Tb, which is the temperature of the skin in the vicinity of the light source. At this point CFT needs to be adjusted to compensate for the increased temperature Tb. Between these two points in time, it is not straightforward to determine the actual temperature of the skin portion, in other words, how much of the skin (typically depth wise) is at Ta, and how much is at Tb. Accordingly, the processing means may provide, at least until steady state conditions are achieved, two values of CFT, one assuming that the tissue is at Ta, and the other correcting this CFT to Tb.

According to another aspect of the invention, measurement of the light intensity is carried out on a skin or tissue portion that is close to but not directly acted upon by the mechanical pressure means. Repeated application of mechanical pressure to the same portion of skin can lead to some minor hemorrhaging of the capillaries in this area, which intensifies the red appearance of this portion. This has the effect of reducing the measured intensity value for the light received therefrom, and thus introduces an error in the determination of CFT. According to this aspect of the invention, the device is adapted for enabling the light or color sensor to receive light reflected from the skin being tested, but not from the part of the skin within this portion that is actually being pressed by the mechanical pressure inducer. In one embodiment, the mechanical pressure inducer is in the form of a plunger, and the light sensor is located above the plunger. In this manner, the plunger itself prevents the part of the skin in contact with the plunger from being visible to the light sensor, which then receives light from the remainder of the skin portion. In another embodiment, the light intensities corresponding to the portion of skin under direct influence from the mechanical pressure inducer is electronically removed from the other light signals. In yet another embodiment, suitable algorithms, embodied in the processing means, disregard all intensity measurements from a predetermined

area of the sensor, corresponding to the area of skin that is subjected to mechanical pressure.

A CFT instrument in accordance with the invention is a non-invasive diagnostic tool which determines the degree to which a patient is in a state of shock, making it possible for a clinician to prescribe a treatment that may save the patient's life. This instrument affords the field of medicine with a new vital sign.

Existing vital signs (pulse rate, respiratory rate, body temperature and often blood pressure) are important signs of life. Also highly significant is a patient's CFT, for this indicates whether a patient is in shock and is in danger of losing his life.

While there has been shown preferred embodiments of CFT instrumentation, it is to be understood that many changes may be made therein without departing from the spirit of the invention.